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# Recording from a Single Motor Unit During Strong Effort

STEEN ANDREASSEN AND ANNELOISE ROSENFALCK

**Abstract**—During strong voluntary effort it is rarely possible to identify the action potentials from single motor units. In large muscles the most selective recordings are obtained with bipolar wire electrodes. To elucidate this experimental finding we have calculated the extracellular field around a single muscle fiber from an intracellular muscle action potential. This model showed that the selectivity of a bipolar electrode is high provided:

- i) the diameter of the recording surfaces is less than half the diameter of the muscle fibers;
- ii) the center distance between the recording surfaces is of the same order or smaller than the diameter of the muscle fibers, and when
- iii) the center-line between the recording surfaces is oriented perpendicular to the direction of the muscle fibers.

A bipolar electrode with these properties will give a maximal attenuation of the field from distant muscle fibers and only two to nine fibers close to the electrode contribute to the recording.

Action potentials were recorded from the anterior tibial muscle of normal subjects with a trifilar cut end electrode and a bifilar side hole electrode. The recording surfaces of the side hole electrode can be positioned close to active muscle fibers by gently pulling the electrode at either side of the muscle.

The recording surfaces should not be smaller than 25  $\mu\text{m}$ . Otherwise the impedances of the recording surfaces are so large that the capacitance between wire and muscle tissue degrades the selectivity of the electrode. Improving selectivity by high-pass filtering was avoided. Narrowing the frequency band made action potentials from different motor units appear similar in shape, thereby making identification more difficult.

The amplitudes and the power spectra of action potentials recorded with these electrodes were in accordance with the model and no more than one to three motor units were present up to 60% of maximal effort.

## INTRODUCTION

THE STUDY of interval patterns of single motor units requires that each action potential throughout a recording period of several seconds can be identified. Bipolar electrodes are best suited for single unit recording from large muscles because they pick up the potentials from the fibers which lie close to the electrodes and attenuate the potentials of distant fibers. During strong voluntary effort it is rarely possible to obtain recordings in which one motor unit is clearly visible without interference from other units [1]–[8]. To clarify whether recording conditions can be improved we have calcu-

lated the potentials from muscle fibers as they would be recorded by unipolar and bipolar electrodes of different dimensions. The extracellular field around a muscle fiber surrounded by an anisotropic conductor was calculated from the intracellular potential according to P. Rosenfalck [9]–[12]. On the basis of this model, we have selected the parameters for two different bipolar electrodes. Finally we have tested the model by recording with the electrodes from the anterior tibial muscle in man.

## MODELING

The properties of a bipolar electrode depend on three factors: i) the *orientation* of the recording surfaces relative to the direction of the muscle fibers, ii) the *distance* between the recording surfaces, and iii) the *areas* of the recording surfaces.

To investigate the dependence on these factors we calculated the extracellular field around a muscle fiber *in situ*. The extracellular potential field was calculated by means of a method suggested by P. Rosenfalck [9]–[12] from an intracellular action potential measured by Ludin [13]. The model takes into account the radius and conduction velocity of the muscle fiber, the anisotropy of muscle tissue, i.e., the difference in conductivity parallel to the fiber direction and perpendicular to it, and the conductivities of the intracellular and extracellular medium (see Appendix).

A unipolar electrode with point-shaped recording surface will record the potential around the fiber. The potential recorded with a bipolar electrode was calculated as the difference in potential between two nearby points. Potentials were calculated for bipolar electrodes lying parallel and perpendicular to the muscle fiber (Fig. 1). If the two recording surfaces have the same distance to the muscle fiber the signal is extinguished (equidistant, Fig. 1).

Potentials were calculated at different distances from a muscle fiber for unipolar electrodes and for bipolar electrodes with a center distance between the recording surfaces of 25  $\mu\text{m}$  (Fig. 2 left).

## RESULTS

In unipolar recording the peak-to-peak amplitude is 1260  $\mu\text{V}$  at distance 0 from the muscle fiber. In bipolar perpendicular recording the potential can be extinguished if the recording surfaces have the same distance to the fiber (equidistant, Fig. 1). With one recording surface at the fiber and the other at distance 25  $\mu\text{m}$  the amplitude is 30% of the potential recorded unipolarly. When the center-line of the bipolar electrode is parallel to the fiber axis the peak-to-peak amplitude is less than 10% of the amplitude in unipolar recording.

With larger center distances ( $c$ ) between the recording sur-

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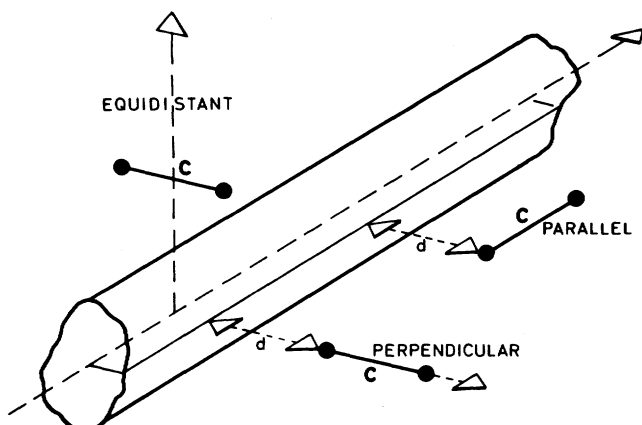


Fig. 1. The two recording surfaces of bipolar electrodes placed relative to the muscle fiber in two orthogonal directions: parallel and perpendicular.  $d$  is the distance from the surface of the muscle fiber to the nearest recording surface, and  $c$  is the center distance between the two recording surfaces.

faces the amplitudes are greater, the increase being about proportional to the center distance when the center-line is parallel to the fiber direction and less steep when it is perpendicular to it (Table 1).

For increasing distance between muscle fiber and electrodes the peak-to-peak amplitude decreases (Fig. 2). To estimate the decline in amplitude as a function of distance between the surface of the fiber and the electrodes, the data are expressed in percent of the maximal amplitude recorded with the electrodes at distance 0 (Fig. 3). The distance at which the amplitude has declined to 25% (25%-distance) is  $116\text{ }\mu\text{m}$  with unipolar recording. With bipolar recording the 25%-distance is 63 and  $76\text{ }\mu\text{m}$  when the electrode is perpendicular and parallel to the fiber direction. The decline in amplitude is two to three times greater for bipolar than for monopolar recording at a distance of  $200\text{ }\mu\text{m}$  and three to five times at  $500\text{ }\mu\text{m}$ . This explains that bipolar electrodes are selective because distant fibers contribute less to the signal in bipolar than in unipolar recording.

The number of fibers which contribute to the signal at maximal effort was estimated by placing a drawing of the pick-up ranges of an electrode with the recording surfaces embedded in a wall of insulating material over a micrograph of the cross-section of an anterior tibial muscle (Fig. 4). The pick-up range is the region within which the center of a muscle fiber should lie if the potential picked up by the electrode should be greater than 25% of the potential from a fiber close to the electrode. With unipolar recording the pick-up range is a semicircle with a radius of  $144\text{ }\mu\text{m}$  (fiber radius:  $28\text{ }\mu\text{m}$  plus the 25%-distance:  $116\text{ }\mu\text{m}$ , Fig. 3). With two recording surfaces placed along the muscle fiber (bipolar, parallel electrode) the pick-up range was  $104\text{ }\mu\text{m}$  ( $28\text{ }\mu\text{m} + 76\text{ }\mu\text{m}$ ). When the two recording surfaces were placed perpendicular to the fibers the pick-up range was considerably smaller because fibers which lie "equidistant" to the recording surfaces do not contribute to the signal. By moving the drawing to different positions over the micrograph the number of fibers within the pick-up ranges were counted as nine to seventeen in unipolar, five to nine in bipolar parallel recording and two to seven in bipolar perpendicular recording.

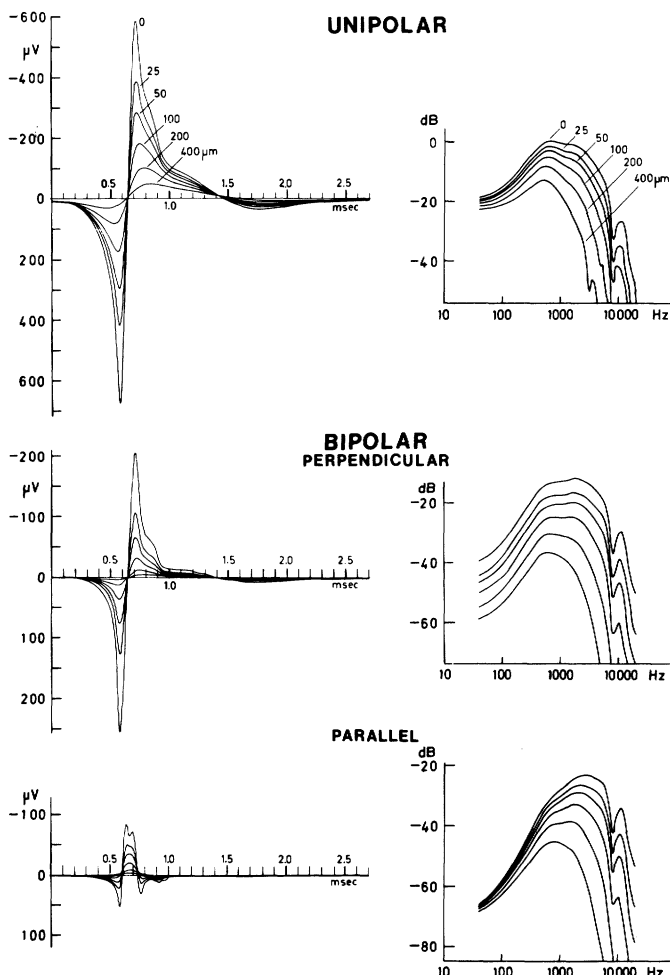


Fig. 2. Left: Action potentials at different distances ( $d$ :  $0\text{ }\mu\text{m}$ ,  $25\text{ }\mu\text{m}$ ,  $50\text{ }\mu\text{m}$ ,  $100\text{ }\mu\text{m}$ ,  $200\text{ }\mu\text{m}$  and  $400\text{ }\mu\text{m}$ ) from a muscle fiber computed from the intracellular action potential for a unipolar electrode (above), for a bipolar electrode with the recording surfaces perpendicular to the fiber axis (middle) and for a bipolar electrode with the recording surfaces parallel to the fiber axis (below). Right: Action potentials at different distances from a muscle fiber converted to the frequency domain by a Fast Fourier Transform.  $0\text{ dB}$  = maximal value of power spectrum for the potential recorded unipolarly at distance 0 from the fiber surface.

TABLE I  
NUMBER OF FIBERS ( $n$ ) WITHIN THE PICK-UP RANGE OF UNIPOLAR AND BIPOLAR ELECTRODES, COMPARED TO THE PEAK-TO-PEAK AMPLITUDES ( $A$ ) OF ACTION POTENTIALS FROM A MUSCLE FIBER CLOSE TO THE ELECTRODES FOR INCREASING DISTANCE ( $c$ ) BETWEEN THE RECORDING SURFACES

| unipolar | bipolar perpendicular  |                        | parallel |
|----------|------------------------|------------------------|----------|
| $n$      | $A\text{ }\mu\text{V}$ | $c\text{ }\mu\text{m}$ | $n$      |
| 9-17     | 1260                   | 25                     | 140      |
|          |                        | 50                     | 250      |
|          |                        | 100                    | 490      |
|          |                        | 200                    | 900      |

The bipolar electrode with recording surfaces placed perpendicular to the fiber direction is thus highly selective when the center distance between the recording surfaces is  $25\text{ }\mu\text{m}$ . With a center distance of  $50\text{ }\mu\text{m}$  the electrode could still be placed

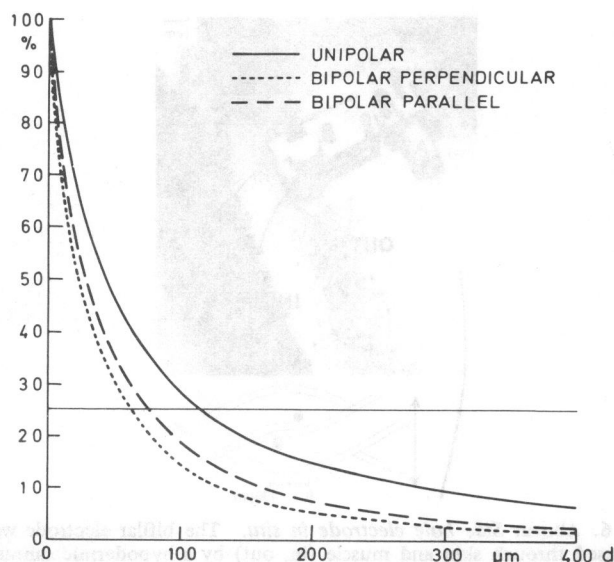


Fig. 3. Decline in peak-to-peak amplitude of the action potential calculated from Fig. 2 for a unipolar electrode and for bipolar electrodes placed perpendicular or parallel to the direction of the muscle fibers. *Ordinate*: Amplitude in per cent of amplitude at distance 0 from the fiber surface. *Abscissa*: Distance  $d$  from fiber surface to the nearest recording surface (see Fig. 1). The distance at which the amplitude has declined to 25% (25%-distance) is determined for the three electrodes by the horizontal line (unipolar 116  $\mu\text{m}$ , bipolar perpendicular 63  $\mu\text{m}$ , bipolar parallel 76  $\mu\text{m}$ ).

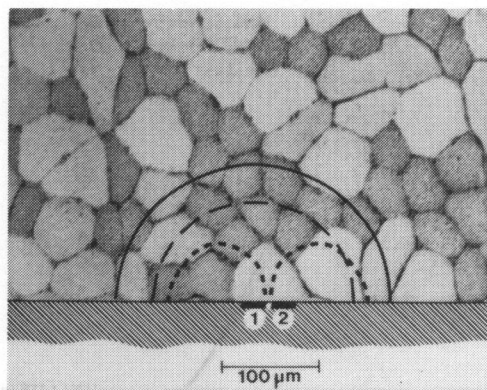


Fig. 4. Pick-up ranges for i) a unipolar electrode (full semicircle) with recording surface at position 1; ii) a bipolar perpendicular electrode (dotted line with two lobes) with recording surfaces at positions 1 and 2; and iii) a bipolar parallel electrode (dotted semicircle) with recording surfaces at position 1 and 25  $\mu\text{m}$  above (or below) position 1. The recording surfaces are embedded in a wall of non-conducting material. The pick-up ranges are superimposed on a micrograph from an anterior tibial muscle in man (courtesy of Dr. Schmalbruch). At various positions on the micrograph the numbers of fibers within the pick-up ranges are nine to seventeen for the unipolar electrode, two to seven for the bipolar perpendicular electrode and five to nine for the bipolar parallel electrode.

such that only two fibers may contribute to the signal but in other positions up to nine fibers may contribute (Table 1). With a center distance of 100  $\mu\text{m}$  the number of fibers within the pick-up range is four to eleven. With center distances greater than 200  $\mu\text{m}$  the pick-up range will be similar to that of two monopolar electrodes. When the bipolar electrode is placed parallel to the fiber direction the pick-up range is independent of the distance between the recording surfaces.

It therefore seems possible to record from few fibers with a bipolar electrode provided the center-line is perpendicular to the fiber axis and the center distance is less than 50  $\mu\text{m}$ . If the center-line is at an angle to the fiber direction the electrode will act as a perpendicular electrode as long as the angle is less than  $45^\circ$ . This is because the perpendicular component of the field at center distances less than 50  $\mu\text{m}$  is three to four times greater than the parallel component of the field (Table 1).

It was assumed in the modeling that the recording surfaces were point-shaped. The calculations are, however, also valid for the 25  $\mu\text{m}$  recording surfaces used in most single fiber electrodes. Using a simpler "tripole" model Ekstedt and Stålberg [14] have shown that recording surfaces which are 25  $\mu\text{m}$  or smaller only slightly decrease the action potential in comparison to recording with point-shaped surfaces.

The shape of the potentials from single muscle fibers depends on the electrode, the duration of the spike being shorter (0.4–0.6 ms) in bipolar than in unipolar recording (1.0 ms). This is important when several motor units are present in a recording because fewer potentials superimpose and each potential from the individual units can thus better be identified.

A quantitative expression for the change in shape with increasing distance between the muscle fiber and the three electrode types is obtained by converting the action potentials in

Fig. 2 left to the frequency domain by a "Fast Fourier Transform" (Fig. 2 right). For zero distance between electrode and fiber surface peak in the power spectrum is at 0.8 kHz for unipolar, at 2.8 kHz for bipolar parallel and at 1.8 kHz for bipolar perpendicular recording. With increasing distance between electrode and fiber the peak moves towards lower frequencies. Note that the upper limiting frequency of the amplifiers and recording equipment should be 7 kHz or better to avoid reduction of the amplitude of the spike potentials from single fibers. From the power spectra the 25%-distances can be calculated as a function of frequency (Fig. 5). The 25%-distance in bipolar perpendicular recording is less than 110  $\mu\text{m}$  through the frequency range, which means that the selectivity is good for all frequencies. This also implies that for the bipolar perpendicular electrode configuration only minor improvements can be obtained by filtering.

#### ELECTRODE DESIGNS

We have used two different wire electrodes, one with side holes and another in which the recording surfaces are the cut ends of the wire.

The *side hole electrode* consisted of two 75  $\mu\text{m}$  stainless steel wires insulated by Teflon and twisted. The recording surfaces were two holes (10–25  $\mu\text{m}$ ) burned into the side of the wire by an induction coil [15], [16].

The side holes were positioned at an angle of  $45^\circ$  to the direction of the wires such that the distance between the recording surfaces perpendicular and parallel to the fiber direction was approximately 50  $\mu\text{m}$  (Fig. 6).

The electrode was inserted by a curved hypodermic cannula which was passed through skin and muscle. The electrode was then pushed forward and the cannula withdrawn (Fig. 6). The recording surfaces were moved close to active muscle fibers by gently pulling on either end of the wire. It could also be

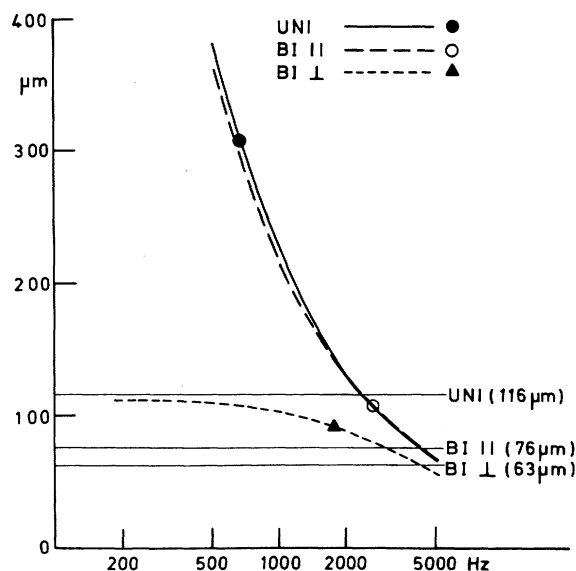


Fig. 5. The 25%-distances for unipolar, bipolar parallel (BI ||) and for bipolar perpendicular (BI ⊥) recording as a function of frequency. The distances were calculated from the power spectra (Fig. 2 right). The horizontal lines indicate the 25%-distance for the peak-to-peak amplitudes of the action potentials as determined in Fig. 3. The peaks in the power spectra are indicated by dots (● unipolar: 0.8 kHz, ○ bipolar parallel: 2.8 kHz, and ▲ bipolar perpendicular: 1.8 kHz).

moved to other positions for further investigations of other motor units.

The cut end electrode consisted of three insulated stainless steel wires, 25  $\mu\text{m}$  in diameter, twisted and glued together with araldite (Fig. 7). The electrode was bent to form a hook and the wire ends were the recording surfaces. The electrode was inserted by a hypodermic needle, which was withdrawn. The orientation of the recording surfaces relative to the fiber direction was random. The action potentials were recorded between the two surfaces which gave single unit responses at the highest level of contraction. According to the model this will be the surfaces which lie perpendicular to the fiber direction. During the first contraction after the electrode was placed into the muscle 5–10 mm of the wire was drawn into the muscle. This “slack” helps the recording surfaces to remain close to the same muscle fiber during contraction. However, once inserted the electrode can only be moved slightly by pressing at the surface of the muscle.

**Electrode holder** (Fig. 6): For both electrodes the thin wires were attached to springs [17]. The cannula used for insertion of the wires was withdrawn and remained on the electrode holder during recording. The holder with the electrode in the cannula was sterilized in boiling water.

## ELECTRODE PROPERTIES

### Electrical Properties

The improvement in selectivity obtained by bipolar perpendicular electrodes relies in part on the shape of the pick-up area, where fibers lying close to the mid-line (Fig. 4) do not give rise to any potential. This only applies if the bipolar electrode records potential difference, or in other words if the electrode preamplifier configuration has a high Common Mode Rejection (CMR). Considering the magnitudes of the unipolar and bipolar potentials (Fig. 2) the CMR should be greater than

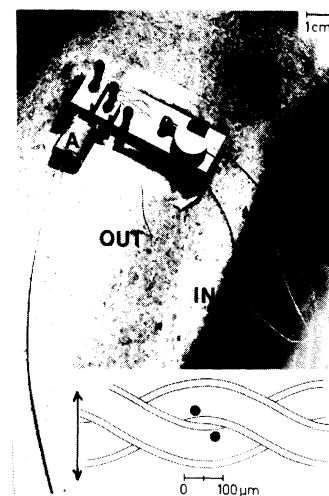


Fig. 6. Above: Side hole electrode in situ. The bifilar electrode was passed through skin and muscle (in, out) by a hypodermic cannula which was withdrawn and remained on the electrode holder during the recording period. The wires were 75  $\mu\text{m}$  stainless steel insulated by Teflon. Below: The recording surfaces were positioned at an angle of  $45^\circ$  to the direction of the muscle fibers (arrow). The center distance perpendicular to the fiber direction was about 50  $\mu\text{m}$ . The wires were fixed to springs on the electrode holder and the input amplifier (A) was plugged directly to these springs. The holder with the springs was sterilized in boiling water.

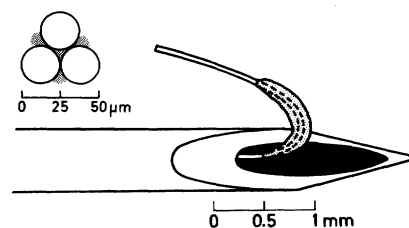


Fig. 7. The cut end electrode was bent to form a hook. It consisted of three 25  $\mu\text{m}$  stainless steel wires insulated by enamel and held together by araldite. The hook was reinforced by araldite to avoid damaging the enamel during insertion of the electrode. Above left: the cut ends of the wires.

ten throughout the frequency range 20 Hz–10 kHz. We measured the parameters that determine CMR, i.e., impedance of the electrode surfaces, capacitance between the electrode wires and the tissue and the preamplifier input impedance (Table 2).

The impedances of the recording surfaces are mainly resistive at frequencies above 1 kHz. They were measured at 3–5 kHz in 0.15% NaCl with a test signal of 1 mV. The impedances of the holes which were burned into the side of the Teflon wire were 160 k $\Omega$ /200 k $\Omega$  in one electrode and 400 k $\Omega$ /700 k $\Omega$  in another. For the cut end electrode the impedances were more similar, 380 k $\Omega$ /400 k $\Omega$ . The impedances were kept low by passing a current 1–10  $\mu\text{A}$  (1–10 s) through them after they were sterilized. The procedure could be repeated with the electrodes in situ if necessary.

The capacitance between the wires and the tissue depends on the thickness and material of the insulation. The side hole electrode was Teflon insulated (2.4 pF/cm) and the cut end electrode enamelled (12.5 pF/cm). When 5 cm of the wires lie in the muscle the capacitances are 12 pF and 60 pF respectively.

The preamplifier was an FET source follower. It was connected directly to the springs on the electrode holder (Fig. 6) to avoid the effect of the capacitances of a shielded input

TABLE 2  
IMPEDANCE OF RECORDING SURFACES ( $R$ ), CAPACITANCE OF THE INPUT  
CIRCUIT ( $C_g$ ) AND THE RESULTING COMMON MODE REJECTION  
(CMR) OF THE INPUT CIRCUIT

|                      | Side hole electrode |   | Cut end electrode   |
|----------------------|---------------------|---|---------------------|
| $R_1/R_2$ k $\Omega$ | 160/200             | 400/700                                     | 380/400             |
| $C_g$ pF             | 20                  | 20  | 70                  |
| CMR                  | > 33 (20–10,000 Hz) | 50 (100 Hz)<br>8 (1,000 Hz)<br>3 (3–10 kHz) | > 25 (20–10,000 Hz) |

$R_1$  and  $R_2$  are the resistive components of the impedances of the two recording surfaces measured at 3–5 kHz in 0.15% NaCl solution (1 mV).  $C_g$  is the sum of the capacitance between the wires and the muscle and the input capacitance of the preamplifier to ground (< 10 pF).

cable which might be of the order of 100 pF. The input impedance was more than 1 G $\Omega$  and the capacitance to ground less than 10 pF.

The Common Mode Rejection (CMR) was measured by applying a common signal to the recording surfaces in 0.15% NaCl. Only 2 mm of the wires around the recording surfaces were dipped. The capacitances between the wires and the tissue were simulated in two ways: i) by connecting two capacitances between the input leads of the preamplifier and ground, or ii) by dipping 5 cm of the wires in a second bath connected to ground. For the cut end electrode less than 4% of the common signal was transferred to the amplifier terminals as a difference signal (CMR > 25). For the side hole electrode with low impedances (160 k $\Omega$ /200 k $\Omega$ ) the CMR was greater than 33. For the side hole electrode with high impedances (400 k $\Omega$ /700 k $\Omega$ ) the CMR decreased inversely proportional to the frequency and reached a level of three at 3 kHz. These measurements of CMR agree with values calculated from the measured values of electrode impedances and wire and amplifier capacitances.

The conclusion from these results is that due to small differences in impedances of the recording surfaces CMR is sufficiently great for the cut end electrode, in spite of the large capacitances. The side hole electrode has smaller capacitances, but a much greater difference between the impedances of the recording surfaces. A calculation shows that the difference in impedances should be kept below 100 k $\Omega$  for the side hole electrode to obtain a CMR greater than ten throughout the frequency range from 20–10,000 Hz. However, since the capacitance between wire and tissue remains constant during the recording period, it may be possible to avoid its limiting effect by using a negative capacitance differential preamplifier.

### Selectivity

Action potentials from single motor units were recorded during constant voluntary contraction of the anterior tibial muscle. The force was measured as torque at the ankle. With the side hole electrode clear recordings from single motor units were obtained at 20–30% of maximal effort over periods of several minutes (Fig. 8). Potentials from one to three motor units could be identified up to 60% of maximal effort. At the highest levels of contraction only few records were stable over more than 30 s. When the side holes were more than 50  $\mu$ m apart the selectivity was much less. In recordings with a multi-electrode, Gath and Stålberg [8] found the optimal distance

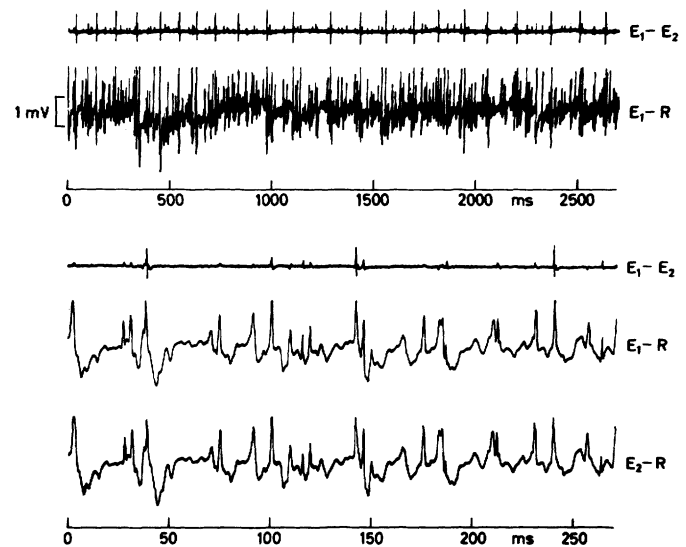


Fig. 8. Clearly visible single-unit potentials recorded bipolarly ( $E_1 - E_2$ ) by a side hole electrode from the anterior tibial muscle at 30% of maximal effort. Upper trace: at low time resolution, third trace: the initial 250 ms at a ten times higher time resolution. The level of activity is illustrated in the other traces recorded between either of the two side holes ( $E_1$  and  $E_2$ ) and a large reference electrode ( $R$ ) placed subcutaneously.

between the recording surfaces to be 140  $\mu$ m. Using 60  $\mu$ m their recording was slightly more selective but very sensitive to electrode movements. Cut end wire electrodes follow the movements of the muscle and can thus stay close to the same muscle fibers at higher levels of contraction. The cut end electrode was very selective and we had the same experience as Clamann and Lamb [7]: in many instances no activity could be obtained in normals at low contraction levels. Recordings from single motor units were obtained up to 60% of maximal effort in normals.

These findings agree with the model. In both electrodes the component of the center distance perpendicular to the fiber direction was less than 50  $\mu$ m and only two to nine fibers are then within the pick-up range of the electrode.

From studies of the fine structure of the motor unit in man it is known that fibers from fifteen to thirty motor units lie intermingled in the larger muscles in man (for review see [18] and [19]). In rat and cat muscle most fibers belonging to the same motor unit lie solitary [20], [21]. Electrophysiological evidence suggests that the fibers of a motor unit also lie scattered in man [22]. In that case about 20% of the two to nine fibers within the pick-up range of the bipolar perpendicular electrode belong to the same motor unit. This illustrates that only in some instances it was possible to record from single motor units at maximal effort as reported by Bigland and Lippold [1], Norris and Gasteiger [2] and Hannerz [5].

### Spectra and Amplitudes of Recorded Potentials

The action potentials were similar in shape and duration to the potentials calculated by the model and there was agreement between power spectra computed from the recorded potentials and determined by the model (Figs. 9 and 2).

The amplitudes of the action potentials ranged from 100  $\mu$ V to 8 mV. With the side hole electrode the median amplitude was 800  $\mu$ V, two times greater than when recorded with the



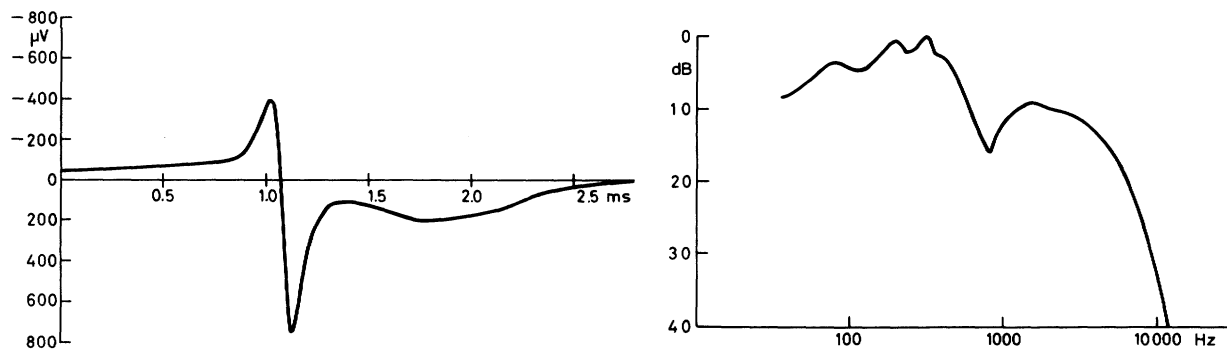


Fig. 9. Action potential recorded by a side hole electrode (left) and its power spectrum (right).

cut end electrode (400  $\mu V$ ). This is primarily because the component of the center distance between the recording surfaces perpendicular to the fiber direction is 50  $\mu m$  in the side hole electrode and 25  $\mu m$  in the cut end electrode. It may also reflect that the side hole electrode can be moved close to the active fibers.

It is surprising that the medians of the amplitudes were 400–800  $\mu V$ , which is close to the maximal amplitudes predicted by the model and that amplitudes as high as 8 mV could occur. To some extent the large amplitudes could be explained by assuming that they were recorded from fibers with diameters larger than 50  $\mu m$ . According to the model, the amplitude of the action potentials increases with the square of the fiber diameter. In addition, the side hole electrode is an insulator except for the small recording surfaces and may thereby cause an increase in the electrical potential by as much as a factor of two (the “wall effect” [23]). The “wall effect” could not play a role when recording with the cut end electrode. To account for the large potentials an additional factor of two or more is needed. We therefore suggest that amplitudes predicted by the model are too low when the gap between muscle fiber and electrode is extremely narrow. The parameters for conductivity and anisotropy were chosen from measurements where the muscle tissue was assumed to be homogeneous. However, the muscle tissue is inhomogeneous and consists of insulating membranes in intra- and extracellular fluid with higher conductivity. The membrane currents flowing during depolarization of the muscle fiber are therefore forced to run in the narrow gaps of extracellular fluid [24]. The current density and thereby the potentials recorded by an electrode in the extracellular fluid immediately outside the fiber are therefore higher than predicted by the model.

In unipolar recording, Ekstedt [23] recorded up to 25 mV with a single fiber multielectrode. The fact that the decline in amplitude for increasing distance between fiber and amplitude was steeper than predicted from the model (Fig. 10) when the maximal amplitude was larger than 10 mV and fitted the theoretical curve for smaller amplitudes, supports the suggestion given above.

#### DISCUSSION

From the model we have learned that when a bipolar electrode of small dimensions is oriented perpendicular to the direction of the muscle fibers only two to seven muscle fibers

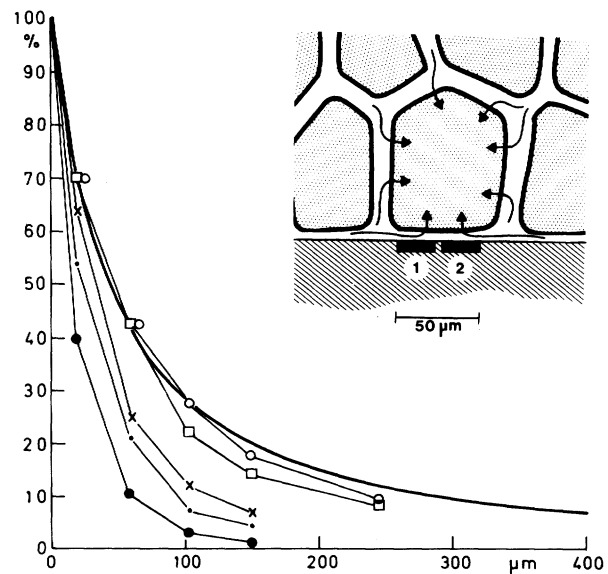


Fig. 10. To explain that the peak-to-peak amplitude recorded with bipolar electrodes can be two to three times greater than predicted by the model. Heavy line: decrease in peak-to-peak amplitude for increasing distance between the surface of the muscle fiber electrode as calculated for unipolar recording (see Fig. 3). The points on the thin lines are calculated from the peak-to-peak amplitudes determined by Ekstedt in experiments with a single-fiber multielectrode [23, Fig. 34]. The values are calculated for a 56  $\mu m$  muscle fiber, taking into account that the abscissa in Ekstedt's figure is distance along the multielectrode. The curves fit the model as long as the amplitudes are 8 mV ( $\circ$ ) and 10.7 mV ( $\square$ ), and deviates for 12.5 mV ( $\bullet$ ), 18.2 mV ( $\times$ ) and 24.5 mV ( $\cdot$ ) indicating that extraordinarily high amplitudes (12–25 mV) can be recorded from single fibers when the fiber is close to the electrode surface. If the gap between fiber and electrode is narrower than the gap between neighboring fibers the current density and thereby the potential in the extracellular fluid at the electrode is increased (as illustrated, upper right) beyond the values predicted from the model. The arrows indicate the current during depolarization.

are within the pick-up range of the electrode. This is in accordance with our experiments: action potentials from no more than two or three motor units were recorded up to 60% of maximal effort.

We have used a trifilar cut end electrode. The pair of recording surfaces perpendicular to the direction of the muscle fibers could be selected when the electrode was *in situ*. The tip of the wires formed a hook which fixed the electrode in the muscle. Once inserted the electrode could not be shifted to other motor units, and if withdrawn the track of damage acted as a



Fig. 11. Above: Selective recording of action potentials from two motor units 1 and 2 via a side hole electrode in the anterior tibial muscle of a normal subject at 20% of maximal effort. Middle: Action potentials from three motor units (1, 2, and 3) selected by differentiation of the lower trace:  $d(E_2 - R)/dt$ . Note the similar shape of the potentials from motor units 1 and 3. Below: Summated activity from many motor units recorded between one of the recording surfaces of the side hole electrode and a reference electrode ( $E_2 - R$ ).

shunt and the selectivity was lost [1]. In patients with partial paralysis who could not recruit all motor units, we used an electrode with holes burnt into the side of the wires. The wires were passed in and out through skin and muscle. The side holes could thus be positioned close to active muscle fibers by gently pulling the ends of the wires at either side of the muscle [6].

Hannerz [5] was able to record action potentials from single motor units from threshold to maximal contraction with a selective wire electrode. It consisted of three silver wires with small holes burnt into the insulation. The best selectivity was obtained when the impedances of the holes were 4 M $\Omega$  and 200 k $\Omega$ . The capacitance between the wire and the muscle tissue plus the capacitance of the input cable and amplifier is presumably above 40 pF. The recording surface with the 4 M $\Omega$  impedance will therefore act as a high-pass filter with cut-off frequency below 1 kHz. Calculations of the transfer function for the bipolar (4 M $\Omega$ /200 k $\Omega$ , 40 pF) electrode show that the common mode rejection of the electrode is low (approximately 1) and that the signal recorded by the electrode is almost identical to the signal recorded unipolarly by the 200 k $\Omega$  recording surface after a differentiation. As the model shows, the selectivity in unipolar recording is improved by high-pass filtering. The recording system used by Hannerz [5] has an upper frequency limit of 1 kHz. The total frequency range of the electrode and recording system is therefore equal to a band-pass filter around 1 kHz. This is below the frequency range where the 25% distances and thereby the pick-up ranges are small (cf. Fig. 5).

Recently the importance of high-pass filtering was emphasized by Gath and Stålberg [8] and Clamann and Lamb [7]. Clamann and Lamb [7] succeeded in extracting single motor unit action potentials by electronic differentiation of the signals picked up by small electrodes. The effect of differentiation of a unipolarly recorded trace is illustrated in Fig. 11. Action potentials from three motor units (1, 2 and 3 in middle trace) were sorted out from an interference pattern (lower trace) by differentiation. The upper trace was recorded from

the same site in the muscle via a side hole electrode. This recording is more selective; the amplitude of the action potentials of motor unit three being below the 25% limit relative to the amplitudes of motor units one and two. This illustrates the relative selectivity of a bipolar parallel and a bipolar perpendicular electrode. The signal recorded by a bipolar parallel electrode is, except for a proportionality factor, identical to the signal obtained by differentiation of a unipolarly recorded signal [25]. This is valid as long as the center distance between recording surfaces is below 100  $\mu$ m.

The middle trace (Fig. 11) illustrates the error which may occur when single potentials are selected by narrowing the frequency band. The potentials from motor units one and three can no longer be distinguished by their amplitude and shape.

In conclusion, we can say that the best selectivity is obtained by recording via a small bipolar electrode, oriented perpendicular to the muscle fiber. Even then single unit recording at maximal effort is only possible when occasionally the electrode is positioned close to two or more fibers belonging to the same motor unit.

#### APPENDIX

The extracellular action potentials  $\phi_{an}(r, z)$  were computed by numerical integration of an intracellular action potential  $V_i$  measured just under the membrane of a human muscle fiber [13], [9]–[12].

For a muscle fiber in an isotropic conductor the extracellular potential was calculated [9] from

$$\phi_e(r, z) = \frac{a^2 \sigma_i}{4\sigma_e} \int_{-\infty}^{\infty} \frac{d^2 V_i / ds^2}{\sqrt{r^2 + (s-z)^2}} ds.$$

$z$  and  $r$  were distances along and perpendicular to the fiber axis respectively.  $\sigma_i$  and  $\sigma_e$  were the conductivities of the intracellular and the extracellular medium,  $a$  the radius of the muscle fiber. The muscle tissue is an anisotropic conductor with larger conductivity  $\sigma_z$  along the fiber than perpendicular to it  $\sigma_r$ . The extracellular potential for a muscle fiber *in situ* was therefore calculated [9] from



$$\phi_{an}(r, z) = \frac{a^2 \sigma_i}{4\sigma_e} \cdot K(z) \cdot \int_{-\infty}^{\infty} \frac{d^2 V_i / ds^2}{\sqrt{r^2 (\sigma_z / \sigma_r) + (s - z)^2}} ds$$

where

$$\phi_e = \sqrt{\sigma_r \cdot \sigma_z}$$

and

$$K(z) = \frac{\delta \phi_e(a, z)}{\delta r} \frac{\delta \phi_e(a \cdot \sqrt{\sigma_z / \sigma_r}, z)}{\delta r}$$

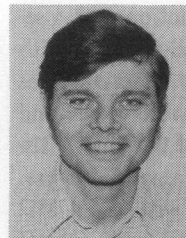
The assumptions were:

- i) radius of the muscle fiber:  $a = 28 \mu\text{m}$
- ii) conduction velocity: 4 m/s
- iii) ratio between intracellular and extracellular conductivity  $\sigma_i / \sigma_e = 1/3$
- iv) Anisotropy:  $\sigma_z / \sigma_r = 2.8$
- v)  $K(z) \cong 1.7$  for  $\sigma_z / \sigma_r \cong 2.8$  [9]
- vi) distance between the centers of the recording surfaces of the bipolar electrodes: 25  $\mu\text{m}$ .

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